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Hand-rim Forces and Gross Mechanical Efficiency at Various Frequencies of Wheelchair Propulsion

Authors

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Key words

- push frequency
- SMART^{Wheel}
- fraction of effective force
- gross efficiency
- wheelchair ergometry

Abstract

To determine the effects of push frequency changes on force application, fraction of effective force (FEF) and gross efficiency (GE) during hand-rim propulsion. 8 male able-bodied participants performed five 4-min sub-maximal exercise bouts at 1.8 m·s⁻¹; the freely chosen frequency (FCF), followed by 4 counter-balanced trials at 60, 80, 120 and 140% FCF. Kinetic data was obtained using a SMART^{Wheel}, measuring forces and moments. The GE was determined as the ratio of external work done and the total energy expended. Increased push frequency led to reductions in peak resultant force ($P < 0.05$), ranging from 167 to 117 N and peak tangential

force ($P < 0.05$), ranging from 117 to 77 N. However, FEF only demonstrated a significant difference between 60% and 140% FCF ($69 \pm 9\%$ and $63 \pm 7\%$, respectively; $P < 0.05$). Work per cycle decreased significantly ($P < 0.05$) and rate of force development increased significantly ($P < 0.05$) with increased push frequency. GE values were significantly lower at 60%, 120% and 140% FCF than 80% and 100% FCF ($P < 0.05$). No meaningful associations were present between FEF and GE. Under the current testing conditions, changes in push frequency are accompanied with changes in the absolute force values, albeit without changes in either the gross pattern/trend of force application or FEF. Changes in GE are not explained by different levels of force effectiveness.

Introduction

A large majority of individuals with spinal cord injuries or lower limb disabilities are dependent upon the use of a manual wheelchair for both daily living and sporting activities. However, the gross efficiency (GE), this being the ratio of the external work done and the total energy expended, of hand-rim propulsion remains somewhat low. Reported GE values range anywhere from 2 to 11% for studies involving able-bodied individuals as well as inexperienced and experienced wheelchair users [13, 17, 30, 34, 36]. In contrast, other forms of upper body locomotion, such as arm cranking [19] and hand-cycling [31] report much greater GE with values commonly ranging from 14 to 19%. The underlying reasons for this remain a topic of interest for research in both rehabilitation and sports environments.

Previous literature has reported GE to be highly influenced by propulsion conditions, such as hand-rim velocity and rolling resistance [30, 34, 37], wheelchair configuration including propulsion mechanism [35], seat height [33], wheel camber

[20], wheel size [21] and differences in motor skills or expertise [10, 11, 17]. Propulsion technique in particular has been shown to be influenced by the push strategy employed; propulsion mode and/or push frequency [8, 17, 18, 38]. These latter studies have found lower arm frequencies to be associated with increased GE yet not always optimised at an individual's self-selected push frequency.

It has been suggested by clinical biomechanists that lower push frequencies are more beneficial than higher frequencies for the health of the musculoskeletal system [3]. The rationale behind this is that lower push frequencies allow for increased push time and a longer push stroke, reducing the number of pushes required per unit of time. Consequently the number of coupling and uncoupling actions of the hand to the hand-rim (as well as the idle recovery phases) is lower as will be the overall segmental (thus muscle) accelerations. There is debate in the literature whether larger forces and moments increase the probability of the risk of injury in wheelchair users [4, 22, 28]. Despite this it would be reasonable to suggest that although the magnitude of force required at lower push frequen-

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cies is greater, the rate of rise of these forces may be reduced as a result of the increase in push time. It has been reported that the rate of force development in wheelchair propulsion is related to the risk of injury [5]. Hence it appears to be beneficial for wheelchair users to: (a) reduce peak hand-rim forces and or push frequency; as well as (b) reduce the rate of rise of force during the push phase of the propulsion cycle to reduce the loading on the joints of the upper body (shoulder, elbow and wrist) involved during propulsion.

When considering the force exerted on the hand-rim it is best described in terms of the radial, axial and tangential components of the resultant (total) force. The radial and axial components create friction between the hand and the hand-rim simultaneously to ensure a tangential force component is applied to the hand-rim [29]. In guided movements, the forces that are applied by the hands do not directly influence the trajectory of the hands. The ratio of the tangential force and the resultant force at the hand-rim gives an indication to what is known in the literature as fraction of effective force (FEF) [30]. The theory of improved FEF from more tangentially directed forces has, however, been disputed [2,9,33]. Efficiency is reduced slightly as a consequence of a learned higher FEF [9]. The concept of FEF and its possible relationship with push frequency and efficiency remains interesting. When mean external work remains constant and push frequency is manipulated then reciprocal changes in the resultant and tangential forces would be anticipated. However, increased push frequencies, above the self-selected frequency, could lead to misdirected tangential forces to a larger extent hence we would report lower FEF at greater frequencies. It is unclear how the ratio of the tangential and resultant forces is affected by push frequency manipulation and whether or not there is an association with push frequency and/or GE.

To our knowledge there is very little literature that has investigated the hand-rim forces during wheelchair hand-rim propulsion under varying conditions of push frequency. Gaining an insight into this type of information should assist our understanding of the relationship of efficiency with push frequency and, extend what is already known in the area. Therefore, the purpose of this study was two-fold; 1) describe the force application profiles of hand-rim propulsion at a range of push frequencies, 2) describe the relationship between force application and GE. We hypothesise that: 1) an increased push frequency reduces absolute force application parameters and FEF; 2) The rate of rise of force increases reciprocally with push frequency and 3) GE decreases with push frequencies that exceed the freely chosen frequency (FCF).

Material and Methods

8 able-bodied male participants (22 ± 4 years) volunteered for this study and gave written informed consent prior to participation following a detailed explanation of all testing procedures. Body mass was recorded to the nearest 0.1 kg using a seated balance scale (Seca 710, Hamburg, Germany) and seated height in the wheelchair was measured to the nearest 0.01 m using a portable height stadiometer. Participants' physical characteristics are given in **Table 1**. Approval for the study procedures was obtained from the University Research Ethics Committee and was conducted in accordance with the Declaration of Helsinki and Ethical Standards in Sport and Exercise Science Research [12]. Participants had prior experimental experience in wheelchair exercise, but were not specifically trained in upper body sports activities or hand-rim wheelchair propulsion.

Table 1 Participant physical characteristics.

Participant	Age (years)	Height (m)	Seated Height (m)	Body Mass (Kg)	Mean PO (W)	Total Resistance (N)
1	20	1.74	1.38	83.8	46.7	26.7
2	20	1.87	1.42	105.1	63.6	35.3
3	19	1.77	1.40	77.8	55.4	31.2
4	24	1.79	1.40	80.9	50.2	27.4
5	31	1.89	1.44	90.9	59.4	33.4
6	19	1.71	1.39	63.3	44.7	24.9
7	21	1.86	1.40	90.8	55.6	30.5
8	22	1.81	1.41	91.5	50.8	28.1
Mean	22	1.81	1.41	85.5	53.3	29.7
SD	4	0.06	0.02	12.3	6.4	3.5

Instrumentation

For the wheelchair trials, all participants were tested in the same 15° cambered hand-rim basketball wheelchair (Quattro, RGK, Burntwood, Staffordshire, England) which was a typical characteristic sports wheelchair used during the early stages of skill acquisition. The wheelchair was configured with a force sensing SMART^{Wheel} (3 Rivers Holdings, Mesa, AZ) to collect kinetic data. Wheels were fitted with the standard solid tyres provided by the SMART^{Wheel} manufacturer (wheel diameter of 0.592-m and hand-rim diameter of 0.534-m). The characteristics and properties of the SMART^{Wheel} are described elsewhere [6,26]. The SMART^{Wheel} was placed on the right side of the wheelchair and its use did not change the camber, axle position or diameter of the basketball wheelchair. To ensure similar inertial properties for the left wheel a counterbalanced weight was added to the wheel. No individual adjustments relative to anthropometrics of the participants were made. The wheelchair was secured to a single roller ergometer (Bromakin; cylinder length, 1.14-m; circumference, 0.48-m). Although velocity was derived from the SMART^{Wheel}, a flywheel sensor was connected to the roller and interfaced to a laptop computer (Compaq Armada 1520, Series 2920A) which was able to calculate and display the wheelchair velocity during trials for participants. Mean power output (Po) was determined from the SMART^{Wheel} and calculated from the torque applied to the wheel axis (Mz) and their angular velocity (ω) [23].

$$\text{Mean Po (W)} = ([\sum (M_z (\text{N}\cdot\text{m}) \cdot \omega (^\circ\cdot\text{s}^{-1}))] \cdot 2) / \text{Samples}$$

As the SMART^{Wheel} measures unilaterally, symmetry was assumed and thus to determine Po the values were multiplied by 2 prior to time averaging to account for work done on the contralateral wheel. The recovery phase was accounted for with Mz (being ≤ 1 Nm) and the angular velocity of the wheel, time averaged from the onset of the first push to the completion of the final push (the end of the recovery phase).

Total resistance was calculated from the mean torque applied to the wheel axis (Mz) and the radius of the wheel as follows:

$$\text{Total Resistance (N)} = [\text{Mean Mz (N}\cdot\text{m)} / \text{Wheel Radius (r)}] \cdot 2$$

Since the wheelchair propulsion was performed at a constant speed the propulsive work done and total resistance must be equal to the resistive work done therefore; it can be assumed that the mean total resistance must be equal to the mean propulsive force which can be calculated.

Testing procedure

The testing followed the same procedure as previously reported experiments [8, 17, 18, 38]. Participants performed a discontinuous, sub-maximal, steady state exercise test on the roller ergometer, consisting of 5 exercise bouts at different push frequencies (FCF and 60%, 80%, 120% and 140% of FCF) at $1.8 \text{ m} \cdot \text{s}^{-1}$. The propulsion velocity employed was selected to ensure sub-maximal exercise for the able-bodied participants based on previous research work [17]. An audio-visual metronome was used to pace the push frequency requirements.

Participants completed a 5-min warm-up prior to performing the sub-maximal push frequency conditions at a self-selected push frequency and propulsion velocity, which was guided with HR not exceeding $130 \text{ beats} \cdot \text{min}^{-1}$. Following an 8-min rest period, a 1-min 'habitation period' was performed to allow the participant to become accustomed with the push frequency to be employed during the following 4-min test period. The FCF condition was the initial 4-min exercise bout and the push frequency was counted and recorded each minute, then the mean frequency was calculated. Subsequent exercise bouts were performed at 60, 80, 120 and 140% of the FCF [17, 18]. An 8-min recovery period separated each test condition to allow for HR to return close to their baseline and permit lactate diffusion. The order of the 4 manipulated exercise bouts was counter-balanced to ensure that each participant performed the conditions in a distinctly different order, thus possible effects of fatigue and/or learning were mitigated.

Kinetic measures

The forces and moments applied to the hand-rim were recorded for 30s during the final minute of each exercise bout. These kinetic data were obtained via an infrared wireless transmitter at 240Hz using the SMART^{Wheel} in the research mode setting. All kinetic data were filtered using the SMART^{Wheel} manufacturer's 32-tap finite impulse response (FIR) low pass digital filter with a cut-off frequency of 20Hz. This process allowed for filtered forces and moments applied for each push frequency to be determined. For each push phase of the propulsion cycle, the SMART^{Wheel} provided the unilateral forces (F) and moments (M) in the 3 wheel-based reference planes, F_x – horizontally forward; F_y – vertically downward; F_z – horizontally inwards; and M_z – referred to the moment produced around the hub in the plane of the wheel [1,6]. The beginning and end of the pushes were derived from the M_z and was identified from the absolute value of 1Nm. The push starts when M_z was $> 1 \text{ Nm}$ and the end of the push was $\leq 1 \text{ Nm}$. The criteria for the push identification was written into a custom excel spread sheet used for processing and analysis of all SMART^{Wheel} data.

The resultant force (F_{RES}), which is the total force applied to the hand-rim, was calculated by vector addition of F_x , F_y and F_z :

$$F_{\text{RES}} = \sqrt{(F_x^2 + F_y^2 + F_z^2)} (\text{N}) \quad [6]$$

The tangential force (F_{TAN}) which is the force directed tangential to the hand-rim, was calculated from torque (M_z) and the hand-rim radius (R_r) and is defined as the ratio between the 2 values, according to:

$$F_{\text{TAN}} = M_z / R_r (\text{N}) \quad [26]$$

The FEF on the hand-rims, by definition the ratio between the magnitude of the resultant force applied and the tangential

component, was calculated for each instant in the measurement period and expressed as a percentage. This method was selected in preference to utilising the ratio between the peak F_{TAN} and Peak F_{RES} as these do not necessarily occur at the same instant.

$$\text{FEF} (\%) = (F_{\text{TAN}} / F_{\text{RES}}) \cdot 100 (\%) \quad [6]$$

The FEF was expressed as the time average FEF over the measurement period. The instantaneous FEFs for each measurement point were time averaged for all complete pushes of the 30s data collection period.

In addition the rate of force development was calculated as the ratio between the changes in F_{RES} from the initial contact to the Peak F_{RES} and the changes in time between these 2 events [4]. All forces and moments were expressed as peak and mean values per push which were then averaged over the total number of pushes produced in the 30s collection period.

Timing

The temporal parameters associated with propulsion were calculated from the kinetic data. Push times (PT) were defined as the amount of time that the hand exerted a positive torque around the wheel axis. Recovery times (RT) were defined as the period of time between the end of a push and the start of the next push. Consequently the cycle time (CT) is the summation of PT and RT. The push angles (PA) were also derived and defined as the relative angle over which the push occurs on the hand-rim.

Physiological measures

Throughout the test, heart rate (HR) was monitored using short-range radio telemetry (PE4000 Polar Sport Tester, Kempele, Finland). Expired air samples were collected and analysed using the Douglas bag technique during the final minute of each condition. The concentrations of oxygen and carbon dioxide in the expired air samples were determined using a paramagnetic oxygen analyser (Series 1400, Servomex Ltd., Sussex, UK) and an infrared carbon dioxide analyser (Series 1400, Servomex Ltd., Sussex, UK). Expired air volumes were measured using a dry gas meter (Harvard Apparatus, Kent, UK) and corrected to standard temperature and pressure (dry). Oxygen uptake ($\dot{V}O_2$) and respiratory exchange ratio (RER) were calculated.

Efficiency

Gross mechanical efficiency was calculated as the ratio of the external work to energy expended during exercise. External work done (W) was determined from the power output (Po) values derived from the SMART^{Wheel} during the hand-rim wheel-chair propulsion for all push frequencies. The metabolic energy expenditure (E) was obtained from the product of $\dot{V}O_2$ and the oxygen energetic equivalent derived from the RER and standard conversion tables [24]. The following equation was used to calculate GE in accordance with previous literature [32]:

$$GE = \frac{W}{E} \cdot 100 (\%)$$

where W is the external work done; E is the total metabolic energy expended.

Statistical analysis

The data were stored and analysed using the Predictive Analytics Software (PASW SPSS for Windows Version 18; SPSS Inc., Chi-

Table 2 Effect of freely chosen push frequency manipulation on kinetic variables.

Kinetic Variables	% Freely Chosen Frequency (FCF)				
	60% (a)	80% (b)	100% (c)	120% (d)	140% (e)
power output (W)	52.6 (5.9)	52.9 (6.6)	54.1 (5.8)	53.7 (7.6)	53.3 (6.7)
push frequency (pushes·min ⁻¹)*	36 (4) ^{b,c,d,e}	48 (6) ^{a,c,d,e}	59 (8) ^{a,b,d,e}	71 (9) ^{a,b,c,e}	83 (11) ^{a,b,c,d}
work per cycle (J)*	83.6 (14.3) ^{b,c,d,e}	65.5 (10.9) ^{a,c,d,e}	55.1 (8.9) ^{a,b,d,e}	45.0 (9.1) ^{a,b,c}	39.8 (8.0) ^{a,b,c}
peak F _{RES} (N)*	168 (31) ^{b,c,d,e}	150 (33) ^{a,d,e}	133 (21) ^a	124 (25) ^{a,b}	118 (23) ^{a,b}
mean F _{RES} (N)*	100 (22) ^{b,c,d,e}	91 (23) ^a	80 (12) ^a	79 (15) ^a	75 (15) ^a
peak F _{TAN} (N)*	117 (20) ^{b,c,d,e}	104 (21) ^{a,d,e}	90 (12) ^a	84 (16) ^{a,b}	78 (15) ^{a,b}
mean F _{TAN} (N)*	70 (16) ^{b,c,d,e}	61 (14) ^a	55 (8) ^a	52 (10) ^a	48 (8) ^a
mean FEF (%)*	69 (9) ^e	66 (8)	68 (5)	65 (8)	63 (7) ^a
rate force development (N·s ⁻¹)*	602 (207) ^{d,e}	684 (278)	669 (181) ^{d,e}	841 (223) ^{a,c}	928 (249) ^{a,c}
push time (s)*	0.38 (0.04) ^{b,c,d,e}	0.34 (0.05) ^{a,d,e}	0.31 (0.03) ^{a,d,e}	0.27 (0.03) ^{a,b,c}	0.27 (0.03) ^{a,b,c}
recovery time (s)*	1.21 (0.19) ^{b,c,d,e}	0.90 (0.15) ^{a,c,d,e}	0.70 (0.12) ^{a,b,d,e}	0.56 (0.10) ^{a,b,c,e}	0.48 (0.09) ^{a,b,c,d}
push angle (°)*	134.7 (11.5) ^{b,c,d,e}	120.3 (13.2) ^{a,d,e}	110.2 (10.6) ^{a,d,e}	96.3 (10.3) ^{a,b,c}	92.6 (11.7) ^{a,b,c}

All values are mean ± SD. * Main Effect of ANOVA

Letters in parentheses after the FCF percentages can be used to identify Bonferroni adjusted pairwise differences in the table

e.g., ^a means the value is different from the 60% FCF condition, whereas ^b is different from the 80% FCF condition and so on

All differences are P ≤ 0.05

cago, USA). Data normality and homogeneity of variance were verified by Shapiro-Wilk and Mauchly's test of sphericity, respectively. The degrees of freedom were adjusted for heterogeneous variances (Greenhouse-Geisser). Standard descriptive statistics (mean ± SD) were calculated for all physiological and kinetic variables. Separate one-way within measures ANOVA were used to examine the effect of the freely chosen push frequency manipulation on kinematic and physiological variables. Bonferroni comparisons were used to identify significant pairwise differences. Relationships between force/timing variables and efficiency/push frequency were examined by Pearson's product moment correlations. A probability threshold of P ≤ 0.05 was considered to be statistically significant.

Results

The 8 able-bodied males physical characteristics are displayed in **Table 1**, with age 22 ± 4 years, height 1.81 ± 0.06 m, seated height 1.41 ± 0.02 m and body mass 85.5 ± 12.3 kg. The participants performed the 5, 4 min exercise bouts with a mean total resistance of 29.7 ± 3.5 N (range 24.9–35.3 N; **Table 1**). The resistance of the wheelchair/roller ergometer system is greater than that of previous literature whereby rolling resistance is generally reported [16–18,35]. Mean power output was 53.3 ± 6.4 W (range 46.7–63.6 W; **Table 1**) due to individual differences in rolling resistance of participants, however, across conditions (60–140% FCF) there was no significant difference in power output (52.6–54.1 W; **Table 2**). The mean FCF was 59 ± 8 pushes·min⁻¹ (**Table 2**). The calculation of the metabolic energy expenditure (used in the calculation of GE) required RER to be ≤ 1.00. However, the maximum energy equivalent of 5.189 kcal (21.7 kJ) was used when the RER for 2 of the participants in the 140% FCF condition exceeded unity (1.00). In this instance, the effect on the GE calculations was deemed to be negligible and separate analysis revealed that removal of these data did not alter the statistical outcome.

The push frequency manipulation had a significant effect on the force application variables (**Table 2**). Peak F_{RES} and peak F_{TAN} declined across the 5 push frequencies; Bonferroni adjusted pairwise comparisons between the push frequencies revealed

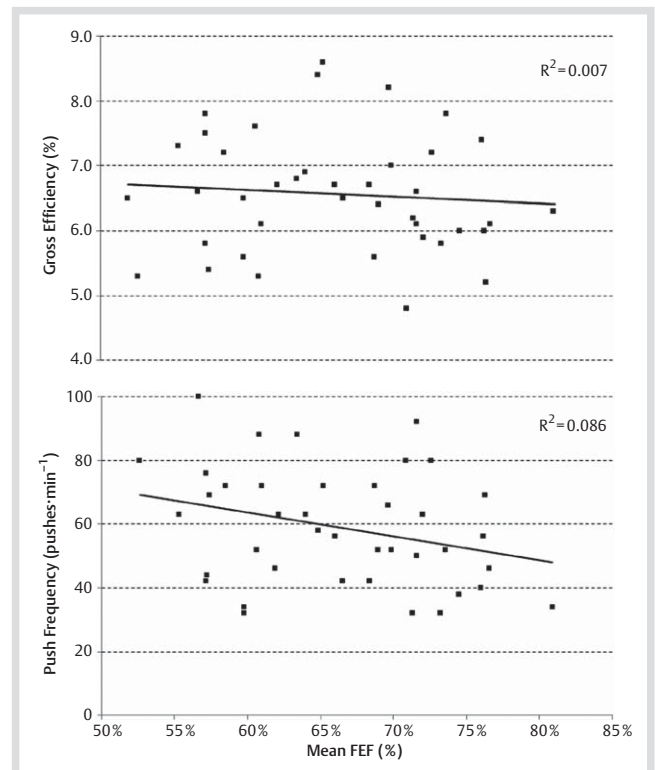


Fig. 1 Relationship between Mean FEF and both gross efficiency and push frequency. R² derived from Pearson Product Moment Correlation.

that the values at 60% FCF were invariably higher than at all of the other frequencies with a more gradual decline from 80% to 140% FCF (**Table 2**). The rate of decline was greatest between the lowest push frequencies of 60–80% FCF and 80–100% FCF (10.7% and 13.5%) in comparison to the higher push frequencies of 100–120% FCF and 120–140% FCF (4.8% and 7.1%). Mean FEF was only significantly different between the 2 extreme push frequencies of 60% and 140% FCF. Mean FEF was not related meaningfully to GE or push frequency (**Fig. 1**). As expected, work per cycle was affected by push frequency (P < 0.05) and decreased with higher push frequency (r = -0.79). Changes in push frequency altered the rate of force development (P < 0.05); rates at

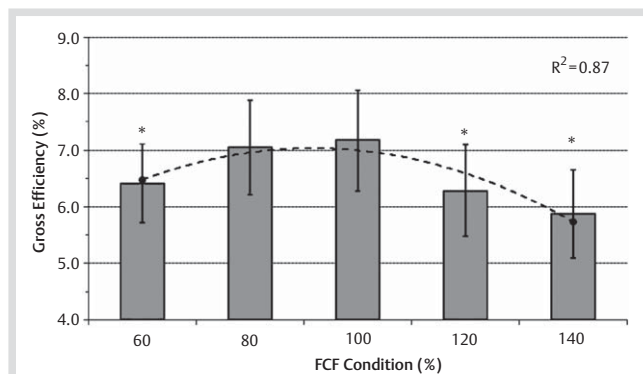


Fig. 2 Gross efficiency values (mean \pm SD) for hand-rim propulsion across range of arm frequencies in the 8 able-bodied participants. * Significant difference between 100% FCF ($P \leq 0.05$). R^2 derived from Pearson Product Moment Correlation.

120% and 140% FCF were significantly higher than at 60% and 100% (• Table 2).

As anticipated, push time, recovery time and the push angle all decreased with increasing push frequency ($P < 0.05$). Push frequency had a significant effect on GE whereby 60%, 120% and 140% FCF were all lower than the FCF (100%; • Fig. 2). As the small difference in GE of 0.1% between 80% and 100% FCF was not significant ($P = 1.00$), it is not possible to conclude that FCF was the most favourable push frequency in our study. The relationship between push frequency and GE appears to be curvilinear with a plateau in GE over the 80% and 100% FCF conditions (• Fig. 2).

Discussion

With a dearth of literature, this study describes the effects of forces applied to the hand-rim during manual wheelchair propulsion whilst specifically manipulating push frequency. This provides an important insight into the association of force application and push frequency, albeit under the experimental conditions imposed. The findings support the first hypothesis that increased push frequency results in the reduction of absolute force for both F_{RES} and F_{TAN} . However, this was not the case with the FEF. The rate of force development increased significantly with increased push frequency supporting the hypothesis of a reciprocal increase with increasing push frequency. The GE showed a curvilinear trend with increased push frequency and supported the hypothesis whereby GE decreases with push frequencies exceeding FCF. There was no association of GE or push frequency with FEF or any of the force parameters. For this reason the effectiveness of force application does not relate to the GE changes observed with changes in push frequency.

Force application

The suggestion that an ineffective force production (low FEF), could in part be responsible for lower GE of propulsion [8, 34] is not supported in this study. Our findings support more recent work that FEF does not correlate to the GE [2, 9, 10, 15]. As de Groot and colleagues [9] have demonstrated, efficiency was lower with a forcefully induced higher FEF compared to a lower FEF. The use of FEF as an indicator of efficient propulsion cannot be supported with the current findings whereby push frequency

was manipulated. It is clear that the most effective propulsion technique from a kinetic/dynamic viewpoint (FEF) is not necessarily the most efficient one. Bregman et al. [2] observed that the force direction during propulsion is a compromise between efficiency and the constraints imposed by the wheelchair-user system. They implied that training should not be aimed at the optimisation of the propulsion force because this may be less efficient and more straining for the musculoskeletal system. Similarly in a recent study on seat height [33] it was reported that simply improving mechanical efficiency through seat height changes does not necessarily optimise the force application characteristics and FEF.

Fraction of effective force across the push frequencies was not reflective of the efficiency. The extreme ends of the push frequency scale (60% and 140% FCF) produced a significant difference in FEF without significant changes in GE. The values of Mean FEF ranged from $63 \pm 7\%$ at 140% FCF to $69 \pm 9\%$ at 60% FCF. These values of FEF are comparable with those found in the literature for hand-rim propulsion in both able-bodied and spinal cord injured participants [2, 7, 33], but slightly lower than the values found by Kotajarvi et al. [15] albeit under different testing conditions. This relatively small and insignificant change in FEF across push frequency may be the direct result of the fact that hand-rim propulsion is a guided movement. Increasing push frequency resulted in lower resultant forces being applied to the hand-rim during propulsion. Interestingly the reductions in peak resultant force were only significant when comparing the 100%, 120% and 140% FCF conditions to the 60% and 80% FCF conditions. This finding was despite the significant changes in push angle and push time throughout the frequency conditions, therefore investigating the rate of force development would appear to be important to help explain the relationship of force with push frequency. As push frequency is manipulated, participants can be seen to adopt a consistent and stable model to satisfy the movement requirements under the given task boundaries of each condition [27]. In each of the different push frequency conditions, the general geometric orientation and co-ordination of the overall upper body remains constant and the arms adapt to the altered frequency by regulating the force magnitude but not its overall effectiveness, i.e. ratio of component forces. The present study indicates that the rate of force development is significantly increased at the higher frequencies. Boninger and colleagues [3, 4] associated increases in cadence, force magnitude and rate of force development with an increased risk of injury. In the context of push frequency it is important to know how this rate of force development is affected and its associated risks. Reduction in push frequency results in a decreased rate of force development although the consequence of this is an increase in peak forces during each push. On the other hand, higher push frequencies demonstrate smaller peak forces but subsequently higher rates of force development more frequently. Therefore, the question remains as to which is better for hand-rim wheelchair propulsion, as results show that the physiological efficiency is not linked to the effectiveness of force application. Push frequency manipulation results in changes to the cyclical timing and the push angles, indicating a reduced push angle with higher push frequencies. Both of these variables are assumed to be responsible for the changes seen in the resultant force applied because of the requirement to maintain the same external workload and thus the production of more or less work per push with a lower or higher push frequency (• Table 2). Hence, at lower arm frequencies there are higher resultant forces

and as push frequency increases the resultant force decreases, however, this resultant force is not significantly reduced statistically within the arm frequency range of 100–140% FCF, although there is a trend for this to continue to decrease and could be clinically significant. The results for peak resultant force and push angle appear to be supported by similar findings in a population of wheelchair users, albeit using a different methodological approach [25]. Richter and colleagues report a self-selected mean cadence of 52 pushes·min⁻¹ along with a decreased cadence (–10%) which are comparable to the push frequencies of the current FCF and 80% FCF conditions. Importantly, they revealed the associated changes in peak resultant force and push angle were of a similar magnitude, although the absolute peak resultant force was much lower [25]. Interestingly this study reports the same relationship and changes in the tangential force; as a result FEF is not affected significantly. Bregman et al. [2] suggested that propulsion technique is mainly determined by the geometrical boundaries of the musculoskeletal system. In that context FEF is suggested to be an invariant characteristic of the biological system, that only changes with extreme geometric changes (i.e., seat height) or with continued learning and training where detailed fine tuning is critical and will lead to (ultra) small long term shifts in FEF. Results of previous studies indeed provide evidence for these notions [2, 9, 10, 15]. Our data support the notion that adaptation to frequency involves a regulation of the force magnitude and movement velocity but does not involve a fundamental shift in co-ordination strategy in this cyclic movement.

Gross efficiency

The present study supports the findings of previous research into the effects of push frequency on GE [8, 17, 18, 38], whereby it has been shown that higher push frequencies (>100% FCF) reduce the GE of propulsion significantly. Unlike previous findings by Lenton et al. [17, 18] the 60% FCF conditions GE was significantly lower, however, it was not possible to identify any significant difference between the 80% and 100% FCF conditions. It is apparent that changes in GE with changes in push frequency are not a direct result of an altered FEF. It seems to be that they must be associated with the different magnitude and frequency of de-/accelerations of the arm segments and trunk, as well as the different ranges in segment excursion and thus muscle contraction velocities, ranges and tension. Different push frequencies result in changes in push angle, thus in the range of motion of the muscles, changes in the force-length/velocity and length tension of the contracting muscles, thus influencing the energy required for the contraction and production of work done against the hand-rim. The increased number of recovery phases, couplings and un-couplings (and the associated small negative braking force increases) increases energy expenditure, elicited from the increased work to move the arms. The role that movement of the trunk and head segment plays during propulsion could offer additional explanation as differences caused by push frequency manipulations may well affect energy cost of the movements. Results of the current study suggest that in the current experimental set-up, push frequencies in untrained subjects at or below FCF are close to optimal energy cost.

Experimental considerations

Able-bodied participants, with limited wheelchair experience, provided a relatively homogenous participant group, not highly trained in any of the push frequencies. Importantly they would be able to perform the exercise conditions sub-maximally, despite the larger power output requirements as a result of the

SMART^{Wheel} use (smaller wheel and increased rolling resistance with solid tyres). Able-bodied participants were used to negate the influence of disability and the probable effects of limited arm or trunk function along with possible effects of habituation and training at given arm frequencies. Use of a standardised basketball wheelchair configuration eliminated any effects of different chair designs/setups, however, it is accepted that this would have an effect on the relative geometry of the chair/user interface for individuals, influencing the physiological demands, force production and propulsion mechanics to that in a conventional daily wheelchair. Nevertheless it is felt that the trends and relationships of the data would not alter significantly. Another consideration was the use of a stationary wheelchair ergometer consisting of a single roller and fixed chain. This was an important feature of the study allowing for the effects of force application in relation to arm frequency manipulation to be investigated without the additional effects of coasting direction (of the wheelchair), and thus the external work requirements. Importantly, the power output across the arm frequencies is equal within each participant. The resistance of the wheelchair/roller ergometer system is greater than previous published literature. The increased resistance could be attributed to a number of factors: Firstly the camber of 15° is significantly greater in the standardised basketball wheelchair than in propulsion studies using everyday wheelchairs. Secondly there is the use of the standard solid tyres provided by the SMARTWheel manufacturer, which have a considerably higher rolling resistance than pneumatic tyres [14, 16]. The third factor is the difference in roller ergometers whereby this study used a single roller ergometer with a much smaller roller circumference than that of the split roller ergometer with significantly greater roller circumference, hence greater resistance. The combination of these factors will have contributed to the higher rolling resistance values reported. The results of this study could be different under different testing conditions, for example; reduced rolling resistance and in different populations of wheelchair users.

Conclusions

▼ In conclusion, increased push frequency generally resulted in a reduction in the absolute values of the force parameters measured and consequently reduced push angle and decreased work/cycle. The exception to this was the rate of force development which increased and FEF remained somewhat unchanged. The FEF and force parameters studied did not reflect the trend in GE of propulsion at different push frequencies, thus supporting current views on FEF suggesting that FEF is invariant under the current testing conditions. Push frequency merely affects the force components in such a way that the ratio of the tangential force to the resultant force remains somewhat proportional to one another despite changes in push angle and push time. Despite the GE of propulsion not being able to be linked directly with FEF it is important to acknowledge the important relationship of force application with push frequency. Results of the current study support previous findings that push frequencies in untrained participants at or below FCF are close to optimal energy cost and efficiency. The practical implications of these results are very important for wheelchair users, coaches or rehabilitation practitioners because they demonstrated the effects that changes in push frequency have on the push forces. Understanding the force changes that occur with changing frequency can support wheelchair users in their choice of push fre-

quency to adopt during daily activities and or sports. Coaches and rehabilitation practitioners may well pay particular attention to the magnitude of the forces and rate of force development for prescription purposes when working with a wheelchair user with respect to the physical capacity of an individual.

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